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Value of biomechanical macromodels as suitable tools for the prevention of work-related low back problems

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Summary

Biomechanical macromodels are evaluated with respect to their possible usefulness for health professionals and ergonomists, as well as for applied research on the prevention of low back problems. It is concluded that in the context stated geometrically simple models, in particular the model by Schultz and co-workers, are to be favoured over more complex models. However, load predictions in extreme trunk postures should be dealt with carefully. It is recommended that the model load predictions should be used only in the comparison of work situations and not for an assessment of the absolute acceptability of a work situation.

Relevance

Low back problems are related to mechanical (over)load at work. This study shows the pros and cons of various biomechanical macromodels as tools for health professionals and ergonomists, as well as for applied research on the prevention of work-related low back problems.

Key words: Low-back pain, work-related, prevention, lumbar, spine, biomechanics, models, evaluation

Introduction

Low back problems and related work absence and disability are a common and expensive problem in western countries. Epidemiological data show a relationship between exposure to mechanical (over)load at work and the incidence rate of low back pain. The exact nature of this relationship, however, still remains obscure.

Occupational health professionals, ergonomists, and applied researchers often have to cope with various issues of the prevention of low back problems, e.g. effects of ergonomic adaptations at the workplace or changes in the work method, acceptability of a work situation (comparison with norms), selection of workers, and return to work of employees after a period of absence. To handle in a proper way the issues of prevention mentioned before, while assuming that low back problems are related to mechanical load on low back structures, the professionals involved should ideally provide answers to one or more of the following questions:

1. Which structures in the low back are supposed to be overloaded (intervertebral disc, vertebra, muscle, or ligament) and under which types of load, e.g. compression force, shear force, or axial torque (i.e. the load criteria)?
2. What is the actual load value for each load criterion in a working situation?
3. What is the maximum acceptable load value for each load criterion (i.e. the norm)?

Generally, compression force on an intervertebral disc (L1-L4, L4-L5, or L5-S1) is chosen as the only load criterion. The lower parts of the lumbar region are chosen because most damage is found to occur in this region. The maximum acceptable value for this load...
criterion can nowadays be obtained experimentally by loading postmortem specimens (following an assumption on the transformation of the results to the in-vivo situation), and in the future possibly by so-called micromodels. Micromodels represent a small part of the spine in detail, usually a functional spinal unit, i.e. two adjacent vertebrae and the intervertebral disc in between.

The actual load value in the work situation can be estimated by a macromodel. Most macromodels consist of two parts, i.e. a so-called free-body diagram and a distribution model. A free-body diagram (see Figure 3, upper part) is used to calculate the forces and moments on the location of interest (here a lower intervertebral disc). This can be done by defining several segments together as a free body (here the upper trunk, the arms, the head, the neck, and a load if present). On the basis of fundamental physical laws the forces and moments, respectively along and around the main axes of the coordinate system, acting on the free body have to be balanced. Here, the forces and moments generated by internal body structures have to counterbalance the forces and moments caused by the so-called external moment (i.e. the weight and position of the various segments included in the free body and various other forces acting on the free body). A distribution model (see Figures 2 and 5) splits these counterbalancing forces and moments into separate contributions of muscle(s), ligament(s), and bony structures.

In this paper published macromodels used within the framework of prevention of work-related low back problems will be evaluated. The evaluation will be focused on the value of the currently available models for applied research, occupational health professionals, and ergonomists. Their needs and interests form the basis of the authors' viewpoints in the evaluation. Therefore the indisputable (potential) value of certain models for increase of insight into low back mechanics will not be so prominent. For a description of future directions of model development and fundamental research the reader is referred to a paper by Chaffin3.

To describe and evaluate the various models of interest we have selected seven characteristics. First, these characteristics and the criteria for evaluation will not be so prominent. For a description of future directions of model development and fundamental research the reader is referred to a paper by Chaffin3.

Table 1. Comparison and evaluation of macromodels on the basis of seven model characteristics

<table>
<thead>
<tr>
<th>Characteristic</th>
<th>Model</th>
</tr>
</thead>
<tbody>
<tr>
<td>1. Geometrical complexity</td>
<td></td>
</tr>
<tr>
<td>肌肉s</td>
<td>M1 Low No ligaments M2a Low No ligaments M2b Moderate High</td>
</tr>
<tr>
<td>韧带</td>
<td>M3 Moderate No ligaments M4 Moderate High</td>
</tr>
<tr>
<td>2. Model techniques</td>
<td></td>
</tr>
<tr>
<td></td>
<td>M1 Correct M2a Correct M2b Correct</td>
</tr>
<tr>
<td>3. Validation</td>
<td></td>
</tr>
<tr>
<td></td>
<td>M1 Good results26 M2a See M1 M2b No adequate results</td>
</tr>
<tr>
<td>4. Muscle force calculation</td>
<td></td>
</tr>
<tr>
<td>Force and moment equations</td>
<td>M1 Force and moment equations M2a Force and moment equations</td>
</tr>
<tr>
<td>Force and moment equations</td>
<td>M2b Force and moment equations + optimization technique</td>
</tr>
<tr>
<td>Force and moment equations</td>
<td>M3 Force and moment equations + optimization technique</td>
</tr>
<tr>
<td>Force and moment equations</td>
<td>M4 Force and moment equations + EMG data</td>
</tr>
<tr>
<td>Force and moment equations</td>
<td>M5 Force and moment equations + optimization technique</td>
</tr>
<tr>
<td>Force and moment equations</td>
<td>M6 Force and moment equations + optimization technique</td>
</tr>
<tr>
<td>5. Static or dynamic input</td>
<td></td>
</tr>
<tr>
<td>Static/ dynamic</td>
<td>M1 Static/ dynamic M2a Static M2b Static</td>
</tr>
<tr>
<td>Dynamic</td>
<td>M3 Static/ dynamic27 M4 Dynamic M5 Dynamic M6 Static</td>
</tr>
<tr>
<td>6. Symmetric or asymmetric input (No. axes in the model)</td>
<td>M1 Symmetric/ asymmetric (1/2) M2a Symmetric/ asymmetric (1/3) M2b Symmetric (1) M3 Asymmetric (3) M4 Asymmetric (3) M5 Symmetric (1) M6 Symmetric for lifting (1)</td>
</tr>
<tr>
<td>7. Acquisition of input data</td>
<td></td>
</tr>
<tr>
<td>Easy</td>
<td>M1 Easy M2a Easy M2b Easy</td>
</tr>
<tr>
<td>Difficult</td>
<td>M3 Easy M4 Easy M5 Easy M6 Difficult</td>
</tr>
</tbody>
</table>

M1 = cantilever models26,29,42-47.
M2 = models by Chaffin et al.: (a) whole-body model74,44-47.
(b) distribution model by Anderson, Chaffin, Herrin, and Matthews69,50.
M3 = model by Schultz et al.6,51-56.
M4 = model by Jäger and Luttmann77-80.
M5 = model by McGill and Norman43,60,61.
M6 = model by Gracovetsky et al.8-10,62-65.
**1. Geometrical complexity**

In general models are used to study a complex relationship in the physical world, i.e. transmission of forces through the back, after a reduction of this complexity to the essential factors (components) and features involved. The central problem is the danger of a reduction leading to neglect of essential parts. On the other hand, increasing the complexity with respect to one group of characteristics, e.g. the geometrical data, should be accompanied generally by a series of related data, e.g. material properties, recruitment patterns of active parts, etc. If such related data are only partially available, the gain of adding more (geometrical) characteristics may be negative in the end. Models of a low level of complexity ('simple models') and of a high level of complexity ('complex models') have both advantages and disadvantages. Simple models do not simulate the exact geometry, e.g. all muscle bundles with their different lines of action. These models use many simplifying approximations. The effects of these approximations are unknown a priori; they can only be estimated. In general the required input parameters to the model can be obtained easily. Complex models allow simulation of more subtle effects. However, it is difficult to obtain all required input parameters. A complex model does not necessarily produce better numerical results than a simple model. Here 'better' means showing more resemblance to the in-vivo situation. In many cases the more global results of simple models are easier to validate and can be interpreted easier than complex models' results.

The geometrical complexity of a macromodel depends largely on its representation of the morphology of muscles and ligaments. A strict requirement for a model is that the structures generating large forces are modelled explicitly. Muscles can deliver force actively, but also passively when stretched. Active muscle forces are more important in normal situations, e.g. at full flexion of the trunk⁴. So a model without muscles (ligaments and thorax (denoted by the concept 'truncal kinematics') are important in all extreme trunk postures, e.g. at full flexion of the trunk⁶. So a model without muscles is useless, and a model without passive forces is of limited use. The actual number of muscles and ligaments required remains questionable.

The relative positions of the pelvis, lumbar vertebrae, and thorax (denoted by the concept 'truncal kinematics') determine the lengths of ligaments and muscles in the lower parts of the trunk, and hence their force and force capacity respectively. Neglecting truncal kinematics in a model may introduce considerable error in extreme trunk postures. Taking truncal kinematics into account makes a model very complex. It will be difficult to obtain accurate input data, as well as data to validate its predictions.

A special phenomenon to be considered in relation to the mechanics of the low back is the so-called intra-abdominal pressure (IAP). The most commonly assumed role of IAP in counterbalancing flexion moments on the low back is a combination of direct loadbearing and the generation of an extending moment. In both ways the compression force on the spine is reduced. IAP may be incorporated in macromodels. However, this does not seem to be of great influence. The compression force on the spine was estimated to vary between a reduction of about 15% on average⁶ and a slight increase⁷ due to IAP. IAP is also supposed to have other roles, such as straining the lumbo-dorsal fascia⁸,⁹ or supporting the spine hydraulically¹⁰. IAP will not be discussed in the macromodel evaluation here, because there are up-to-date reviews on this topic¹¹,¹².

Concerning the evaluation of the models' geometrical complexity the authors take a view that a complex model should justify its use instead of a simple model. In case of model predictions on the same load criterion, the predictions of the complex model should deviate from the simple model's predictions. It should be recognized that this standpoint holds only within the framework of model use by occupational health professionals, ergonomists, and applied researchers as a practical tool to prevent work-related low back problems. This implies that a complex model can still increase insight into the mechanics of the low back without producing outcomes deviating from a simple model. Predictions from complex models on load criteria not given by simple models due to their character are a justification in itself for the use of the complex model. Apart from this the validity of all model predictions, either from a simple or from a complex model, has to be shown (cf. characteristic 3).

**2. Model techniques**

With respect to model techniques a methodological part can be discerned from a numerical part, i.e. the way in which the equations of a model are solved. Although the numerical part often has a considerable influence on the outcome and value of the model, it cannot be evaluated properly on the basis of its description in a paper. Therefore only the methodological part of the models will be discussed.

First of all the model descriptions should be clear, correct, and unambiguous. In general, descriptions have to be present on the assumptions at the root of the calculation procedures and the calculation procedures themselves (e.g. force-body diagrams, force and moment equations), as well as on the values of input parameters. The modelling of, for example, muscles and ligaments with respect to cross-sectional areas, lengths, lines of action, attachment points, and maximum tension should be described in detail.

**3. Validation**

Validation is a crucial issue in the use of mathematical models. After all, a mathematical model is only a set of equations. Its link to reality is through the physical properties of the system it is meant to simulate¹³. Validation is defined as the comparison of a model prediction and a measurement result on the same
parameter or related parameters. In the latter case a strong relationship between both parameters should be demonstrable. Also the criteria for comparison should be given. If a model is to be used for the comparison of conditions (inputs, e.g. work situations, working postures) a linear relationship with or without an intercept difference between predicted and measured values (showing the same trend at certain conditions) is sufficient. For purposes of determination of absolute acceptability of a certain condition an exact one-to-one relationship is necessary. Ideally, validation experiments should be done for a wide variety of conditions within the scope of applicability of the model. Within the framework of prevention of work-related low back problems, validation on a parameter related to this issue is essential.

Validation is particularly important in complex models containing many phenomenological relationships. Macromodels calculate forces in ligaments, muscles, and joints, the degree of detail depending on the model. Except for some cases of invasive research in general these forces either are not or cannot be measured directly in vivo, due to ethical or technical reasons. Active muscle forces can be estimated in a semidirect way from electromyography (EMG). However, the conversion factor (gain factor) between electrical and mechanical activity varies among different positions of electrodes on a muscle and among muscles. If force predictions are validated using EMG, it is implicitly assumed that passive forces are negligible, or at least constant. This can be assumed to be true in non-extreme and in static postures only. Passive forces (from muscles and ligaments) might at best be approximated by estimating the strain of the structures (length minus resting length × 100) and calculating the stress (force per unit area) with the aid of a (known) stress–strain relationship. From the calculated stress and the cross-sectional area the force can be calculated.

The statements made in this subsection so far reflect a pure scientific viewpoint. It is imaginable that for policy-making less strict requirements are accepted. In that case at least sufficient confidence should still be presented on the correctness and the scope of applicability of model predictions. Confidence on the correctness of model predictions can be presented by carrying out sensitivity analyses. For example, ligament strain depends heavily on truncal kinematics. Therefore, the effects of estimated measurement error and true variation with respect to this factor should be calculated in order to get insight into the confidence interval for the ligament strain predicted.

4. Muscle force calculation

The human body usually has more than one way of distributing the total amount of required counterbalancing forces (and moments) over the available active muscle components and the passive structures (ligaments and passive muscle components). The forces delivered by passive structures depend directly on their strain, stress–strain relationship, and cross-sectional area. The remaining force to be counterbalanced has to be delivered by muscles.

In the low back for the counterbalance of each force (or moment) more than one muscle is available. The maximum force of a muscle is determined by its physiological cross-sectional area, its maximum force per unit area, its strain, and its strain rate (strain per unit time). The actual force delivered by a muscle depends on the activity of its nerves.

The strategy the body follows in force (and moment) distribution is not known. This means that in a model of the low back the system is not determined. An assumption must be followed in order to be able to calculate the forces of the individual muscles. Three approaches are commonly used:

1. Muscles are grouped in functional units till the system is determined. Actually this is equivalent to a simulation with fewer muscles.
2. The activity of muscles in a model is prescribed using a measured EMG signal. The major flaw of this method is that in general it is impossible to convert a measured EMG signal to a muscle force, except for special static conditions. The use of EMG therefore requires assumptions on gain factors, i.e. the relation between measured EMG and the muscle force (mV/N).
3. The muscle forces are calculated by optimization techniques. Often the technique of linear programming is used. A linear programme is an optimization model that can be stated in the following form:

minimize $\sum_{j=1}^{n} c_j x_j$ (the objective function)

subject to: $\sum_{j=1}^{n} a_{ij} x_j \geq b_i$, $i=1,2,...,m$ (the constraints)

where $c_j$, $b_i$ and $a_{ij}$ are the known parameters and $x_j$ are the unknown variables. The known parameters include moment arm lengths, line-of-action orientations, muscle cross-sectional areas, and external moments. The unknown variables are the joint reaction and muscle forces.

The problem of the distribution of forces over individual muscles around a joint is discussed vividly. Of special interest within the framework of this study is a study on the influence of the physiological cross-sectional area of muscles around the hip. It emerged that the calculated forces for the individual muscles vary much more (a factor 2 to 8) than the calculated joint compression force (11%). This suggests that forces of individual muscles are much more sensitive to the optimization approach chosen than joint compression forces.
It is the authors' view that grouping of muscles (approach 1) should not be preferred in solving the indeterminate problem. In all conceivable loading conditions, except for pure flexion moments, the geometrically complex arrangement of trunk muscles causes undeniably counterbalancing forces and moments around two or three main axes of the coordinate system. This reality is violated by choosing approach 1. Furthermore, measured EMG signals (approach 2) are also not preferred, leaving their use for validation. The remarks above lead to a preference for an optimization technique to take away the indeterminate problem.

5. Static or dynamic input

Neglecting inertial effects, as occurs in static modelling, may cause considerable error in calculated loads.24,25 Moreover, the ability to deal with dynamic loading enlarges the scope of applicability of the model. Any model can be made quasidynamic by including inertial forces of the load and the upper body.

6. Symmetric or asymmetric input (number of axes in the model)

During labour there is hardly any situation which loads the body in a symmetrical way. As in the previous model characteristic, the ability to deal with asymmetric loads and trunk postures enlarges the scope of applicability of the model considerably.

A one-axial model has the ability to handle loads around one major axis of a local coordinate system at the low back, e.g. the frontal, the sagittal, or the longitudinal axis. Two- or three-axial models can handle (combined) loads around two or three of the axes mentioned before.

Taking a closer look at the model characteristics static/dynamic (cf. characteristic 5) and symmetric/asymmetric, it is felt that the ability of a model to handle asymmetric loads should be given a higher value than the ability to deal with (quasi)dynamic loads. The reason for this distinction is the fact that making a model able to handle asymmetric loads requires a major change of the distribution model of the low back. The process of accurate data acquisition gets more difficult while making a model able to handle quasidynamic loads as well as while making a model able to handle asymmetric loads.

7. Acquisition of input data

The process of acquiring input data gets more difficult by the number and the type of input variables required. This is important for the user of the model, especially in field situations.

Description of the macromodels

Six macromodels have been distinguished. The first macromodel, i.e. a cantilever model, has been described by many authors. The other five macromodels
come from research groups around one or two central authors. Table 1 compares the macromodels according to the characteristics mentioned in the previous section. The models are presented in sequence of increasing geometrical complexity. As the free-body diagrams of most macromodels are rather similar, the description will be focused mainly on the distribution models.

Cantilever models (M1)\textsuperscript{25,26,29--42} describe the spine as a cantilever in which the external moment is counterbalanced by increased tension in an activator at the back, representing all force-producing structures (or its analogue at the ventral or lateral side of the trunk). One-axial cantilever models contain only one activator (Figures 1 and 2), two-axial models (Figure 3), two activators. There is no distribution problem, because the number of unknown activator forces equals the number of moment equations. Some cantilever models have been validated by comparing, for instance, calculated forces and measured EMG\textsuperscript{26}.

The model by Chaffin and co-workers (M2a)\textsuperscript{28,44--47} was used originally to determine the load capacity of the whole body. For the low back the model incorporates a simple cantilever model. The body consists of ten rigid segments (Figure 4). The positions of the feet and the load at the hands are prescribed. The relative positions of the segments can be varied by the model in such a way that the loading moment on any major body joint does not exceed its muscular strength, and that the compression force on the functional spinal unit L\textsubscript{5}--S\textsubscript{1} does not go beyond a maximum accepted value. The model was used to
formulate guidelines for maximum forces to be exerted by hand for any hand position relative to the position of the feet. The maximum hand force exerted in a seated position as predicted by the model shows a strong correlation with the maximum hand force measured. This kind of validation is only acceptable for this type of model, but not for a distribution model. The model can be considered to have had a major influence on the NIOSH guidelines for manual lifting of loads.

Later versions of the model are user-friendly, 2D and 3D computer programmes to calculate muscle load to load capacity at the major body joints as well as the compression force on the low back for a given body posture and force on the hands.

The sagittal plane distribution model by Anderson, Chaffin, Herrin, and Matthews (M2b) includes two extensor muscles (erector spinae muscle and multifidus muscle) and various ligaments around the L5–S1 functional spinal unit. L5–S1 kinematics are used to predict ligament strains. The model was validated by comparing model predictions on intradiscal pressure and on intra-abdominal pressure with measurement results from literature. In a minority of cases the same trend was observed for measurement results and model predictions. This concerned a rise in intradiscal pressure for increased trunk flexion as well as for increased load in the hands. Within the framework of model use as a practical tool for prevention of low back problems, this is considered not convincing for a moderately complex model. The benefits from using the model instead of a simple model are not visible. Furthermore, within the framework mentioned the use of intra-abdominal pressure for validation purposes is questionable. The relationship between intra-abdominal pressure and the load on the lower lumbar spine may be rather weak.

The three-dimensional distribution model by Schultz and co-workers (M3) mainly contains five pairs of bilaterally arranged muscles (Figure 5). The force in each muscle is calculated using linear programming. Several objective functions have been used, e.g. minimum spine compression, and minimum muscle contraction intensity. The model is validated for the group of test subjects by comparing quantitatively calculated muscle force and measured EMG (for the erector spinae muscle(s) the average EMG from four skin electrodes was taken). The correlation coefficients for the erector spinae muscles were high for sagittal plane trunk postures and various types of asymmetric loading (r > 0.91), but lower for more complex asymmetric trunk postures and loading (0.60 < r < 0.88). Next the model is validated by comparing quantitatively calculated spine compression force and measured intradiscal pressure (r = 0.94).

This model was followed by models with seven and eleven pairs of bilaterally arranged muscles, both having a more refined erector spinae representation, the latter including the psoas and quadratus lumborum muscles. These more complex models turned out to be as valid as the model with five pairs of bilaterally arranged muscles.

The three-dimensional distribution model by Jäger and Luttmann (M4) contains one rectus abdominal muscle and a pair of bilaterally arranged erector spinae muscles. Next to this, two oblique abdominal muscles, representing the left internal and right external oblique muscles and vice versa, are incorporated. Individual muscle forces are calculated by an optimization procedure, minimizing the sum of all muscle forces required.

The distribution model by McGill and Norman (M5) incorporates extensive anatomical detail of the three-dimensional musculoligamentous skeletal system. Muscular activity is prescribed by a measured EMG signal. Truncal kinematics is included in the model. The data (force per unit area of muscles, stress–strain relationships of ligaments, etc.) used in the simulations are described in the thesis by McGill. The model is very complex.

The distribution model by Gracovetsky and co-workers (M6) also incorporates extensive anatomical detail. The description of the model is, however, very scanty. No free-body diagrams are presented. Many details of the model are omitted, e.g. stress–strain relationships, moment arms and
lengths of ligaments, as well as force per unit area of muscles. The model seems to be extremely complex. Predicted back muscle forces were compared with EMG measurements from literature for lifting up to 27 kg with a 40° bent back. Although a resemblance, i.e. the same trend, was found, this cannot be considered a validation for this model as a whole with its enormous pretensions. The model postulates a subtle relationship between the role of the ligamentous system and the roles of the muscles. This relationship depends on the amount of trunk flexion and related truncal kinematics, and the amount of load lifted up to about 200 kilogram. Furthermore, two roles of IAP, i.e. by functioning as a hydraulic amplifier and by straining the lumbodorsal fascia, are described to get the ligamentous system under active muscle control. None of the predictions related to these essentials of the model has been validated.

**Evaluation of the models**

A comparison and evaluation of the models based upon the model characteristics and criteria described before is given in Table 1. In this section first the model characteristic number 1, geometrical complexity, will be evaluated by comparing the models. Secondly, more information will be presented on the evaluation with respect to the model characteristics 2–4, model techniques, validation, and muscle force calculation respectively. Thirdly, the evaluation on model characteristics 5–7, static versus dynamic input, symmetric versus asymmetric input, and acquisition of input data respectively, will be elucidated. The last three characteristics can be grouped under the heading ‘practical usefulness’.

**Geometrical complexity**

With regard to the geometrical complexity of the macromodels, it can be seen that the cantilever models, the whole-body model by Chaffin and co-workers, the model by Schultz and co-workers, and the model by Jäger and Luttmann do not describe the complexity of the back in enough detail to provide more insight into its mechanics. Contrary to the more complex models, the simple models are insensitive for truncal kinematics, which determines the lengths of muscles and ligaments and hence their force capacity and force respectively.

In the opinion of the authors the use of complex models has to be justified by experimental results, or at least sufficient confidence, showing that predictions not given by simple models or predictions deviating from simple models’ predictions on the same load criterion are valid.

None of the moderately or highly complex models studied fulfils the requirement of presenting experimental results or by providing sufficient confidence on the validity of model predictions that are additional to the predictions from simple models. This concerns mainly the predictions on ligament strain. No confidence intervals are presented based on a sensitivity analysis on the effects of estimated (measurement) error and true variation with respect to the factors in the models that determine ligament strain.

The sole load criterion available for comparison of models is the predicted compression force on lower lumbar discs under sagittal plane loading conditions, as only for this criterion could enough data be found in the literature. In a cantilever model and in the model by Jäger and Luttmann the erector spinae muscle(s) generates 100% of the counterbalancing moment; in the model by Schultz and co-workers this is roughly estimated at 90% (remainder by the latissimus dorsi muscle); in the model by Anderson, Chaffin, Herrin, and Matthews around 80% (including the multifidus muscle); and in the model by McGill and Norman 75–80% (sacrospinalis muscle), leaving aside the possible role of intra-abdominal pressure in all these models. The compression force on the lower lumbar discs is determined largely by these muscular forces. Compression force is also dependent on the muscle moment arm(s) used in a model. According to McGill and Norman almost no differences in calculated compression forces existed between their complex model and a cantilever model using an extensor muscle moment arm of 7.5 cm (instead of the previously often used 5 cm moment arm). It can be assumed that an (average) moment arm of 7.5 cm is more realistic, due to its basis in extensive anatomical study. This notion leads to two conclusions. First, an anatomically complex model served a basic scientific purpose by providing relevant modelling data, but for model application purposes within the framework discussed in this study a simple model with a 7.5 cm extensor muscle moment arm can be used instead. Second, modelling uncertainty with respect to muscle moment arm is considerably reduced, i.e. to a small range around 7.5 cm.

The importance of the magnitude of the numerical differences among the models shown above (muscle force and moment arm contributions, and their effect on compression force) can be seen by comparison with measured compression forces in vivo during sagittal plane loading conditions. Nachemson has determined that the compression force on a lower lumbar disc (L₃–L₄) equals about 700 N during upright standing, approximately 2500 N while holding 40 kg with a 20° bent back and straight knees. The magnitude of compression force measured during various activities can thus be reckoned to vary by a factor of 3–5. This means that the differences in calculated compression forces among the models are relatively small as compared to the differences occurring among different loading conditions. So there is no ground to express a strong preference for one particular model.

On the basis of the reasoning before in this subsection it is concluded that for loading conditions in the sagittal plane the use of a simple cantilever model
(including the whole-body model by Chaffin et al.) is sufficient to assist a health professional in comparing, for example, the effects on low back load of ergonomic adaptations at a workplace or the differences between work methods (e.g. lifting techniques). This holds also for the Schultz et al. model that acts like a cantilever model under these loading conditions.

**Model techniques, validation, and muscle force calculation**

The major flaw of the model by Gracovetsky and co-workers is its incomplete description, which makes the model rather useless for a thorough and fair evaluation. Many of Gracovetsky's ideas might prove to be interesting, if presented adequately.

The McGill and Norman model has to be credited for the inclusion of an increased number of anatomical details on muscle lines of action and moment arms. As a consequence in simple cantilever models an extensor muscle moment arm of 7.5 cm can be used instead of the previously often used 5 cm moment arm. Also the model raises two points of criticism. Firstly, EMG signals from only six electrodes are used to drive as many as 36 muscles in the model. Secondly, the model uses a common gain factor (error term) to correct all individual (EMG-based) predicted muscle forces in such a way that their moment sum balances the external moment. From the descriptions and data presented it can be deduced that in the dynamic situations studied this common gain factor varies by a factor of 2 during the period of high loading, meaning that a large discrepancy exists between the muscle forces required to balance the external moment and the predicted muscle forces. This discrepancy most certainly reflects a complex of problems accompanying the determination of dynamic muscle forces on the basis of a dynamic EMG and a static EMG–muscle force relationship.

A major advantage of the model by Schultz et al. as opposed to all other models described is its moderate to good score on validation experiments for a wide variety of (a)symmetric loading conditions and upper body postures seen most during physical labour.

**Static or dynamic input, symmetric or asymmetric input, and acquisition of input data**

At last we come to the three model characteristics related to the practical usefulness of the various models. Two-axial cantilever models (Figure 3) can to a certain extent deal with some asymmetric postures of arms and trunk flexion/extension, and lateral flexion. The Schultz et al. model and the Jäger and Luttmann model can be used for the same purposes, but in addition are able to deal with axial rotation loads, with or without axially rotated trunk postures. The advantage of the more recent versions of the whole-body model by Chaffin et al. is that its low back cantilever model is included in a whole-body biomechanical model. This whole-body model determines the muscle load to strength ratio on most of the major body joints in addition to the calculation of the low back load. Furthermore, it is included in a user-friendly computer program.

All models need input on the posture of the major body segments (arms, neck, trunk, etc.). Some of them can handle dynamic movements. Model predictions in extreme trunk postures should be dealt with carefully because of the likely increased contribution of passive forces and their as yet limited validation.

The model by McGill and Norman has limited potential for practical use. Because it is rather cumbersome to get the EMG input needed, the model can be used under laboratory conditions, and possibly in some real work environments. Large-scale application by health professionals is certainly not realistic.

**Discussion**

In the introduction several issues were raised which are to be dealt with by health professionals, ergonomists, and applied researchers. In terms of quantifying the relative effects on the back load of ergonomic adaptations at the workplace and changes of the work method, the use of any simple macromodel will do, as stated in the evaluation of the models' geometrical complexity.

To deal with issues related to the absolute acceptability of a work situation, however, i.e. a comparison with certain norms, apart from a macromodel norms are also needed. In these cases the exactitude of the absolute load value calculated by a macromodel is very important. The authors of this article doubt whether the absolute load values predicted by the current macromodels are valid. For example compression forces on functional spinal units strongly depend on the muscle moment arm(s) chosen. Let us assume that during trunk flexion the major part of the counter-balancing moment is delivered by the erector spinae muscle (or sacrospinalis muscle). A moment arm of 7.5 cm instead of the more commonly used 5 cm yields a 33% decrease for the component of the compression force generated by the muscle, which is by and large the most important component.

Of no less importance to the issues related to the absolute acceptability of a work situation is the validity of the norms used. These norms are derived from loading experiments on postmortem specimens. The maximum acceptable load values found show a large variation among the various studies and specimens used. This means that to provide a safety norm for a large percentile of the (worker) population (e.g. 95th or 99th percentile) the norm has to be set so low that hardly any loading situation during work is allowed, even with very conservative macromodel load predictions. Future research with respect to norms should at least focus on explanatory parameters for the current large variation in load capacity (e.g. compressive strength) between specimens. Brinckmann...
et al.\textsuperscript{68} have related the compressive strength of a functional spinal unit to the product of bone mineral content and the surface area of its vertebral endplate. The two independent variables may be determined \textit{in vivo}, directly or by some intermediate relationship. This opens an interesting perspective for the comparison and the possible translation of the postmortem results in the \textit{in vivo} situation.

It is concluded that at this juncture a determination of the absolute acceptability of a working situation with respect to its effects on the low back by comparing a macromodel load prediction to a norm derived from experimental data on postmortem specimens cannot be recommended. The authors feel that a promising alternative way is offered by large-scale epidemiological research. In this approach model load predictions on the same denominator(s), i.e. any load criterion, in various work situations are related to the prevalence and severity of low back problems.\textsuperscript{2,3} On the basis of results a norm can be set or adapted.

Conclusions

1. The use of simple macromodels is sufficient for comparing low back load predictions for sagittal plane loading conditions.
2. The macromodel by Schultz and co-workers is to be favoured above others due to its simplicity, its three-dimensional character, and its moderate to good results on validation experiments for various (a)symmetric loading conditions and trunk postures.
3. Macromodel load predictions in extreme trunk postures have to be dealt with carefully.

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